

## Degradation-Induced Transmission Losses in Silica Optical Fibers

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**Background and Objective:** The interaction of surgical optical fibers with tissue has been studied.

**Study Design/Methods and Materials:** Fibers (600  $\mu\text{m}$ ) were lased in chicken and beef tissue using a Nd:YAG laser from 5 to 50 W in both cw and pulsed modes.

**Results:** With longer lasing and higher power, larger transmission loss and degradation (burn-in) of the fiber tip, occurred. This degradation converts the Nd:YAG laser power to heat and leads to further energy loss. During contact lasing, tissue and blood adhere to the fiber tip surface limiting laser transmission, desiccating, and eventually destroying adhering tissues. Such tissue residues create high power densities and temperatures at the tip, which then cause a variety of degradation processes to be initiated.

**Conclusion:** "Burned-in" fibers do not photocoagulate; rather they incise tissue. With continued lasing, thermal shock, chemical, and mechanical breakdown of the fiber leads to failure of the fiber tip and the spalling of glass fragments into the tissue bed. *Lasers Surg. Med.* 21:65–71, 1997. © 1997 Wiley-Liss, Inc.

**Key words:** optical fibers; laser surgery; Nd:YAG laser; silica fibers; degradation; photocoagulation

### INTRODUCTION

In recent years, there have been many investigations of the laser-tissue interaction [1–4]. However, little research has been devoted to studying the response of the optical fiber material (silica glass) when it interacts with tissue [5–8]. In a noncontact mode, the optical fiber works well, capable of transmitting > 90% of the laser light. In this mode, the laser functions as a photo coagulator.

In a contact mode, however, laser surgeons routinely "burn" the fiber in by contacting either surrounding tissue or a wooden tongue blade. Such burn-in is necessary to enable the fiber to function in an incision mode. Once burned in, the laser surgeon can use the fiber only for tissue incision and/or ablation. Even so, many clinical benefits are found using such fiber in the contact mode [9–13].

Although the burn-in process is critical for

the fiber to work in the contact incision mode, almost no work has been performed to study this process in detail. Studies that do exist concentrate on the response of the tissue to a burned-in fiber rather than studying the fiber itself and the changes that must obviously take place to change the nature of the fiber-tissue interaction. It is the purpose of this report to provide the first detailed investigation of the optical fiber before, during, and after the burn-in process.

Figure 1 shows the effect of contact Nd:YAG laser surgery on a typical silica optical fiber. As clearly seen, the damage is extensive and for this

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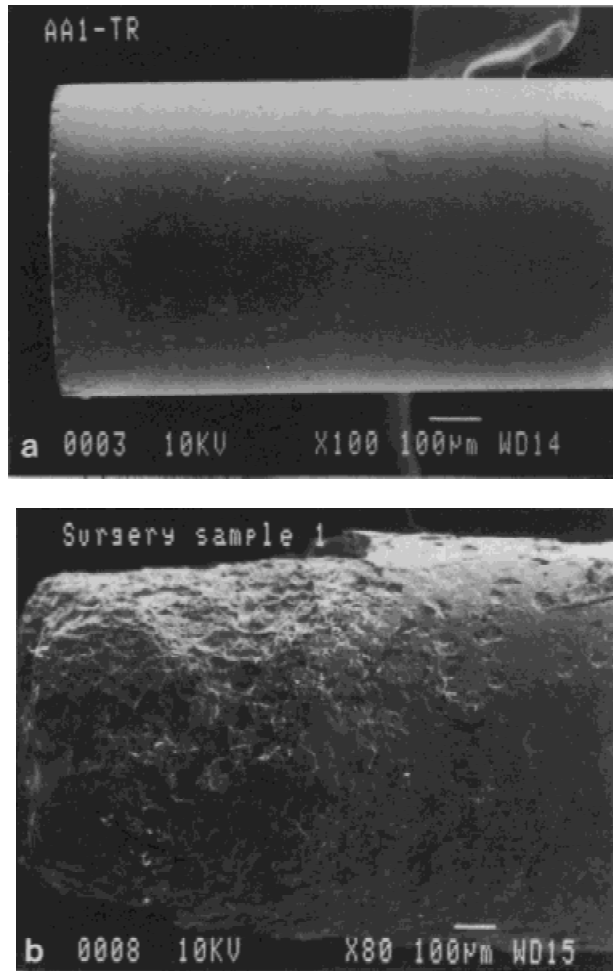


Fig. 1. (a) SEM micrograph of pristine silica optical fiber tip. (b) SEM micrograph of a silica optical fiber tip after a typical gynecological surgery.

reason is expected to have a profound effect on the performance of the fiber during laser surgery. This degradation and its effect on the laser power transmission of the optical fiber are in detail in this report. Before reporting our results, however, it is important to review the current literature on this problem.

Dowlathshahi et al. [14] reported the reduction in laser power transmission after fiber damage and observed that when heated, tissue adheres to the fiber tip and causes strong absorption of the laser energy resulting in temperatures sufficient to melt the fiber. Some authors have attributed the fiber damage to the buildup of heat [9, 14–28] in which the charring of blood insulates the fiber tip, thereby raising the temperature. Rapid heat dissipation into the surrounding tissues is prevented due to the changes in the tis-

sue after photo coagulation [19, 20]. It has been suggested that carbon from desiccated tissues causes the intense heat that then damages the fiber tip [15, 21–25]. Deposited carbon at the tip will absorb most of the energy from the laser and cause intense, highly localized heating of the fiber tip. Verdaasdonk et al. [22] reported that adherent carbonized tissue particles caused absorption of ~33% of the laser light in hemispherical optical fiber probes causing temperatures on the probe tip to reach ~1,000°C. It was also observed that the adherent carbonized tissue particles could not be removed even during rigorous cleaning processes, thereby indicating permanent fiber damage. Isner et al. [26] suggested that defects or imperfections in the fiber tip from mechanical polishing may also induce mild focusing of the laser beam within the fiber tip resulting in absorption and hence fiber tip heat up.

In all of this, an investigation of the optical fiber interaction during laser surgery is still lacking. The purpose of this work is to investigate the interaction between tissues and silica optical fibers by examining the materials aspect of the burn-in process and the resulting optical fiber tip degradation.

## MATERIALS AND METHODS

### Target Tissue

Deboned and deskinmed chicken breasts and beef liver were used as the target tissues. During the experiments, the tissues were kept moist by flushing with deionized water or saline solution. The surgery simulation consisted in scanning the silica fiber tip across the tissues at ~1 cm/sec., a speed that approximates that commonly used by laser surgeons. Experiments were also conducted by holding the fiber perpendicular to the tissue surface and imbedded to a depth of a few millimeters.

### Laser

The Nd:YAG laser used in this study is a Cooper Lasersonics 6000 operated in cw or pulsed modes (2 sec on and 2 sec off) at powers from 5 to 50 W as measured out the distal end of the optical fiber. Some loss in laser power, ~3%, was observed with these fibers and was assumed to arise from their multimode nature.

### Optical Fibers

Flexible blunt-end silica fibers (typical style B36D, Lasersonics) 2–3 m in length were used to

deliver the laser energy to the target tissue. These are step index fibers with a 600- $\mu\text{m}$  diameter core clad with an acrylic resin and protected with a  $\sim 0.5$  mm thick Teflon<sup>TM</sup> jacket. To polish the blunt-end silica tips, a Cooper fiber repair kit (model #8750) was used, which consisted of a series of polishing papers, down to 6  $\mu\text{m}$ .

#### Power Meter

A computer interface power meter (model DGX, Ophir) was used and power readings were made every 0.5 sec for a total of 60 readings. During the measurements, the fibers were placed  $\sim 1$  cm away from the detector.

#### Scanning Electron Microscopy

A JSM-840A SEM was used to investigate the effect of contact lasing on the microstructure of the fiber tips. The fiber tip samples were mounted on 1" carbon stubs and sputtered with gold to a thickness of  $\sim 200\text{\AA}$ . The SEM was operated at an accelerating voltage between 5–15 keV using a working distance of 15 or 30 mm. Magnifications used in this study ranged from 95 to 500 $\times$ .

#### Lasing Studies

Numerous experiments were performed using various laser power levels and lasing times. Most of the experiments were conducted by lasing the tissue with a polished blunt-end silica fiber either in the perpendicular static or scanning modes until burn-in occurred. Burn-in is visually determined by the burning of the target tissue. The burned-in fiber is then scanned over the saline moisten tissue at the preset power and lasing time. After each experiment,  $\sim 2$  cm of the fiber tip was cleaved and labeled for further analysis.

#### Power Transmission Studies

The power transmission studies consisted of measuring the laser power transmission out the fiber tip at a set power level of the laser before and after the lasing experiment. The power levels varied from 10–20 W; the lasing time varied from burn-in to 5 minutes of continuous lasing.

#### Burn-in Studies

The fiber was placed perpendicular to the target tissue until burn-in occurred. Burn-in was noted by a flash of light at the fiber tip and burning of the target tissue. The power levels varied from 5–50 W.

#### Continuous Lasing Studies

Continuous lasing studies consisted of scanning ( $\sim 1$  cm/sec) the target tissue for a set time after burn-in had occurred. The scanning times varied from 15 seconds to 5 minutes for both cw and pulsed laser modes; the power varied from 5–50 W.

### RESULTS

#### Power Transmission Results

Figure 2 shows the transmission loss as a function of laser power and time after the fiber tip had been burned-in. For each power level, time equal zero corresponds to the time when the fiber tip had just burned-in and under most conditions was shorter, the higher the laser power setting. At 10 W, there is little measurable transmission loss just after burn-in. However, there are considerable laser power losses after burn-in at 15 and 20 W. As the lasing time increases, most of the fibers showed a systematic increase in the loss of power transmission. For example, at 20 W after lasing for 5 minutes in chicken, the fiber had a 47% decrease in power transmission, net 53% laser power transmission out the fiber tip end. At higher powers, more heat is generated thereby accelerating and enhancing the degradation of the silica fibers. Hence, longer lasing times and higher powers drastically reduced the forward power transmission of the laser light.

#### Burn-in Results

As shown in Fig. 1, the surface of the silica fiber before surgery is smooth with sharp, well-defined distal end edges. Figure 3 shows silica fiber tips immediately after burn-in at 15W (a) and 20 W (b) in chicken, respectively. Although morphological damage is minor; there are blister-like formations appearing near the distal end of the tips.

#### Continuous Lasing

As the fiber tip is lased in the tissue after burn-in, the tip undergoes further degradation. Figure 4 shows micrographs of silica fiber tips that have been lased in chicken after burn-in for 1 minute and 2 minutes, respectively, at 15 W. Morphological damage is evident in the fiber tips lased in the target tissue. Cracks are now being initiated with more blisterlike formations, and flaking occurs. Damage appears to be increasing as the lasing time increases, i.e., the fiber tip

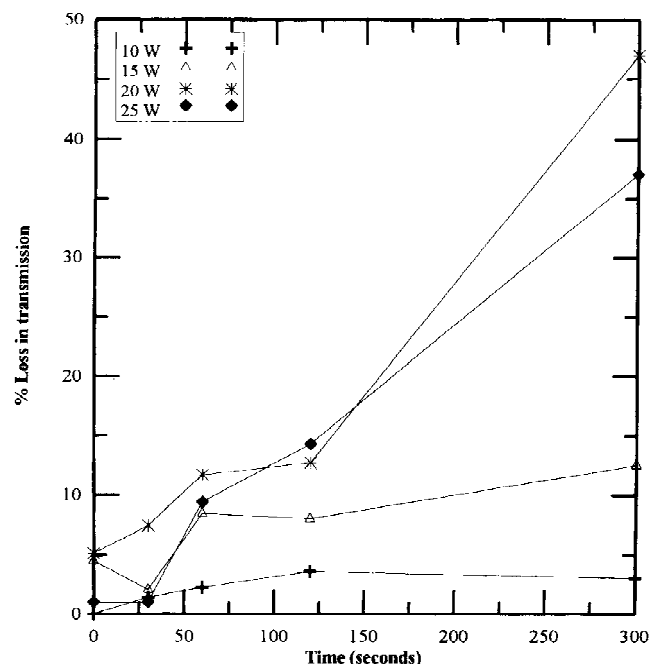


Fig. 2. Laser power transmission loss out the fiber tip end as a function of laser power and lasing time after burn-in.

lased for 2 minutes sustains more damage than the fiber tip lased for 1 minute. Figure 5 shows a silica fiber tip after being pulsed in liver for 2 seconds on and 2 seconds off for 5 minutes at 15 W. The fiber tip shows catastrophic damage. The pulsed mode fiber tip test better simulates the lasing conditions used in laser surgery since the laser surgeon frequently turns on and off the laser power throughout the surgical procedure. The similar appearance of the laboratory fiber tip (Fig. 5), and the surgical fiber tip (Fig. 1b) reflects the similarity in both tissue and lasing conditions between the two procedures. The liver tissue on average also is thought to better simulate human tissue due to its increased vascularity.

## DISCUSSION

Contact of tissue with a bare blunt-end silica optical fiber results in not only tissue desiccation, but also fiber damage. As long as the fiber tip does not contact tissue, transmission of laser power remains high, small near zero power transmission losses occur, and the laser light will photo-coagulate tissue. When contact is made, the fiber tip becomes damaged and the amount of laser power transmitted out the fiber tip is severely reduced. The process of damaging the tip has been termed

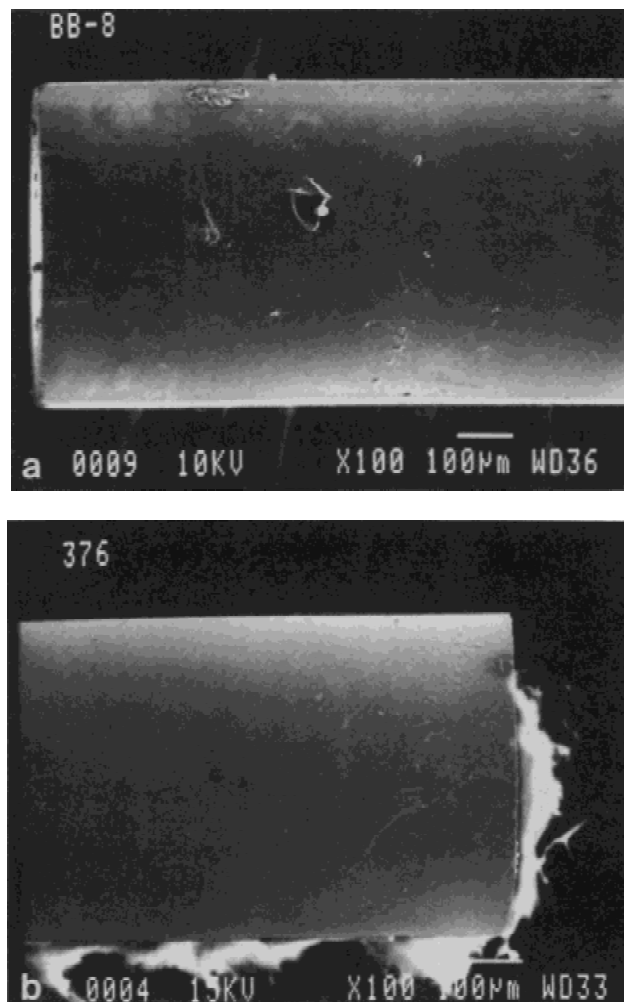


Fig. 3. (a) SEM micrograph of a silica optical fiber after burn-in in chicken breast at 15 W. (b) SEM micrograph of a silica optical fiber tip after burn-in in chicken breast at 20 W.

as "burn-in." Frequently, when a fiber is burned-in, a loud popping sound is emitted with a simultaneous flash of light indicating possible optical breakdown and oxidation of carbon. The result after burn-in is a hot fiber tip that incises tissue.

From the current results, a preliminary description of fiber tip degradation in contact with tissue can be hypothesized. A pristine fiber emits the 1064 nm wavelength laser light when connected to the Nd:YAG laser. Cells absorb this energy, begin to dehydrate, and become sticky and adhere to each other at temperatures ~ 40–50°C. When the fiber contacts this tissue, the cells will then adhere to the fiber tip. At temperatures from 60–100°C, photo-coagulation occurs. Tissue buildup on the tip will absorb the laser energy, causing the temperature at the tip/tissue inter-



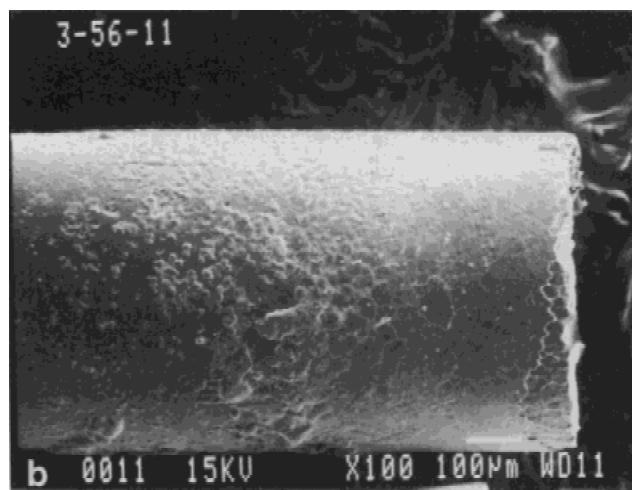
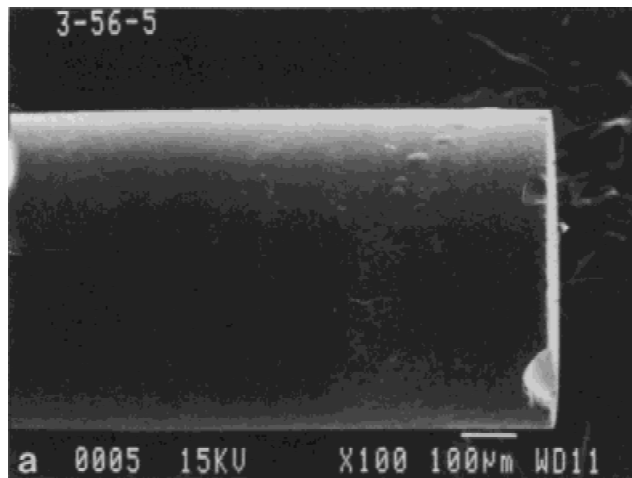


Fig. 4. (a) SEM micrograph of a silica optical fiber after lased in chicken breast for 1 minute at 20 W. (b) SEM micrograph of a silica optical fiber after lased in chicken breast for 2 minutes at 20 W.

face to increase until tissue desiccation occurs. Tissue desiccation leaves a layer of carbonaceous material on the fiber tip. Carbon will strongly absorb the laser light and oxidize, creating a large temperature spike, estimated to be  $>900^{\circ}\text{C}$  at the tip. This temperature rise causes the surface of the fiber to be highly stressed, leading to development of severe thermal shock stresses as well as blistering and melting of the fiber tip. This temperature rise also is believed to oxidize the optical fiber's acrylic cladding. Once the cladding has been burned off the fiber tip, the optical fiber will cease to function as an optical light guide since the water/tissue interface against the fiber will naturally not act as an optical cladding. The fiber is now burned-in. It will no longer photocoag-

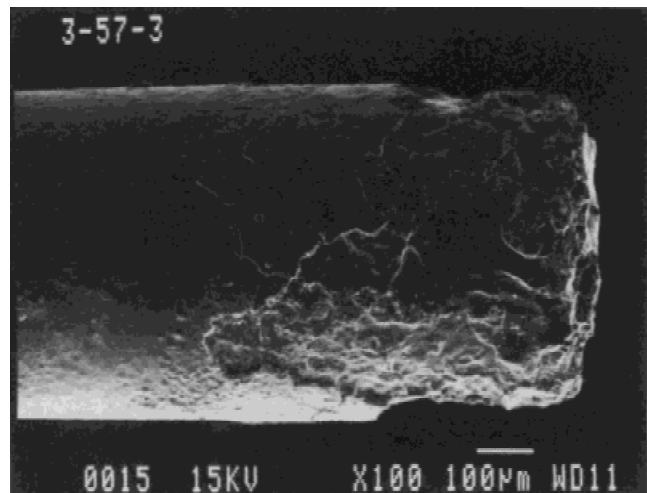


Fig. 5. SEM micrograph of a silica optical fiber pulsed in liver (2 seconds on and 2 seconds off) for 5 minutes at 15 W.

ulate, but rather incise tissue. Visually, the end and  $\sim 1$  mm of the side of the burned-in fiber tip is free of any debris, but farther back along the side of the fiber, carbonaceous material and the burnt back cladding is observed.

SEM micrographs of fiber tips immediately after burn-in show no major fractures in the tip surface, rather only the formation of blisters are observed. Hence, the extensive damage fibers as shown in Figures 1 and 5 are the result of continued use of the fiber tip beyond the point of burn-in. The increasing power transmission losses for these fibers as shown in Figure 2 then correlates with this proposed increasing extent of thermal mechanical breakdown of the fiber tip. The fiber tip temperature is also observed to increase with increased damage, and this must be a result of the increased absorption of laser power by the fiber tip. Hence, the incision ability of the fiber tip is a direct result of the fiber tip thermal mechanical degradation. Such behavior suggests that the current laser surgical fiber technology is not well optimized since tissue incision depends intimately and apparently solely on degrading the very fiber the surgeon is using.

Although it is commonly believed that silica glass is a very chemically inert material, it is well known and characterized to the contrary that silica glass is a very chemically active glass, especially in aqueous systems. It is known, e.g., that pristine silica glass surfaces rapidly hydrate to form siloxyl groups at the glass surface. Furthermore, silica glass is also quite porous on an atomic

scale and molecular water can actively diffuse into silica glass where along with alkali and alkaline earth ions naturally occurring in interstitial tissue fluids, they attack both the fiber surface and bulk glass to form structure degrading nonbridging oxygens. Such chemical degradation of the silica fiber tip must be occurring simultaneous to and accelerated by any thermal mechanical degradation of the fiber tip. Indeed, as the silica glass fiber tip becomes either hydrated with terminal siloxyl groups or alkali and/or alkaline earth bearing nonbridging oxygens, the mechanical and thermal properties of the glass are dramatically degraded. For example, the softening point of silica decreases  $\sim 1,000^{\circ}\text{C}$  by adding only 10% of alkali or alkaline earth nonbridging oxygens. Simultaneous to this, the thermal expansion coefficient increases by more than an order of magnitude and in so doing dramatically decreases the thermal shock resistance of the glass. It is therefore reasonable to propose that significant degradation of the structure and hence thermal-mechanical properties of the glass fiber result not only due to the heat up of the fiber tip as described above, but also from the chemical attack described here. In order to control the degradation of the fiber tip, it will necessary to control both of these degradation mechanisms.

Along with the chemical attack, it is hypothesized that intrinsic changes within the fiber also occur, and one result of this is that light waves can propagate out of the sides of the fiber when contacting tissue. During burn-in, a temperature gradient may form between the inside and outside of the fiber. As the surface tip temperature increases so does its volume, causing a volume differential between the interior and exterior of the fiber. The volume difference between the inside and outside of the tip could cause microcracks, thermal shock failure as described above, that reflect and scatter the laser light. Also, the volume change stresses the bonding structure of the glass thereby leading to increased transmission losses. Also, such bond stresses create a index of refraction gradient within the fiber, thereby resulting increased scattering of the laser light.

The burnt cladding and carbonaceous material observed farther back from the fiber tip also may play a role in the burn-in process. Any debris adhering to the bare glass fiber core will induce laser waves, termed evanescent waves, to propagate out of the glass fiber core into this debris, thereby heating up the debris as well as the fiber. Evanescent waves naturally escape into a

normal cladding, but decay exponentially if the core, cladding, and core/cladding interface are homogeneous and uniform; therefore, transmission is not greatly effected. Problems arise, however, if the core/cladding interface is inhomogeneous, or the cladding is removed thereby exposing the core. Any debris that contacts the core will result in an inhomogeneous boundary layer and induce the evanescent waves to propagate out of the core.

SEM micrographs of continuously lased or pulsed fiber tips show extensive damage to the tip and along the sides of the fiber. From this it is proposed that evanescent waves propagate out the sides of the fiber where adhering tissue debris absorb the laser power and create high power densities and hence high temperatures. Thermally induced breakdown of the chemical structure of the silica glass fiber then occurs.

Finally, there are several other mechanisms that can further degrade the structure and surface of the fiber tip. These mechanisms include explosive vaporization of water and mechanical wear. Direct contact between the moist tissues and a hot fiber tip may lead to explosive boiling. After burn-in, the fiber tips can be extremely hot ( $>1,000^{\circ}\text{C}$ ), which can cause water vapor bubble growth in the moist tissue to become explosive, e.g., rapidly forming vapor can cause physical explosions. The resulting shock wave caused by the violent explosions could damage the tip. In addition, the explosions can generate high impact pieces of tissue, tissue debris, or even glass fragments themselves to impinge upon the tip. From these thermally induced effects, i.e., dissociation and ionization of the vaporized material and production of shock waves, the fiber tip degrades. The many explosions due to vaporization and the resulting shock waves bombard against the fiber tip and cause chipping, cracking, and pitting of its surface. Chips are the sloughed off due to mechanical wear as the fiber tip is scanned along the tissue surface.

Clearly, such dramatic degradation of the laser optical fiber during surgery is not an advantageous process. In order to tailor the optical fiber more carefully to the needs of the laser surgical procedure, this degradation process must be controlled and eventually eliminated. It is to this latter problem that our research is now directed. In a future publication, a new laser optical fiber that has been optimized to withstand the rigors of the laser surgery as described above will be described.

## CONCLUSIONS

From the results presented, silica optical fibers undergo burn-in, spalling, and cracking during laser surgery. High temperatures and power densities generated during laser surgery degrades the chemical makeup the silica fiber, resulting in scattering and absorption of the laser light. These effected dramatically change the functionality of the laser optical fiber. Prior to burn-in, the fiber acts as a photocoagulator. After burn-in, the fiber acts a hot tip incisor as the laser power at the fiber tip is converted to heat.

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